A FULLY NONLINEAR FINITE ELEMENT MODEL FOR ORTHODONTIC TOOTH MOVEMENT PREDICTION.

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1. ABSTRACT

This work presents a finite element model of a tooth and its surrounding tissues in orthodontic tooth movements. It includes the following features: the periodontal ligament mechanical behaviour is modelled as nonlinear bilateral contact conditions between the bone and the tooth; the alveolar bone is considered as an anisotropic organisation of elastoplastic trabeculae; a remodelling algorithm accounting for the pressure dependency due to the periodontal ligament cellular activity is used. We test this model for two types of loading. The first, applying a pressure to the crown, leads to a remodelling along the whole height of the tooth, leading to a tipping movement. The displacement of the tooth for a constant force increases in time due to remodelling. The second, applying a constant rotation of the tooth, shows a decrease of the force needed to keep the rotation angle as well as a latency period before a decrease of the force. Using a pressure dependency of the remodelling rate therefore allows representing most features of orthodontic tooth movement in both the loading cases presented.

2. INTRODUCTION

Orthodontic tooth movement (OTM) is the result of bone remodelling at the interface with the periodontal ligament (PDL) around a mechanically loaded tooth in response to a biomechanical stimulus. A simulation-based treatment using computer models could allow accounting for some patient-specific features in the treatment planning, and therefore OTM software would be a valuable tool for the orthodontist. Such a simulation tool needs to address several issues such as acquiring the geometry of both the anatomical features and the appliances but also the (bio-)mechanical behaviour of all involved tissues and materials. We will address in this work two issues: first, the periodontal ligament representation in patient-specific models; then we will present a fully integrated remodelling and constitutive non linear model for the alveolar bone.

For the past few years, 3D finite element (FE) models based on computer tomography (CT) are increasingly used in the field of OTM [1,2 among many others]. However, clinical CT resolution allows only for differentiation of bone and teeth. Specially, the surface geometry of the PDL cannot be directly derived from CT images. In most recent studies, the PDL is generated using scaling and/or Boolean operations on the teeth and bone interface in order to obtain a thin enclosure [1,2]. This approximation is performed despite the fact that most authors agree on the importance of geometrical and material properties of the PDL in the achievement of OTM. We here present a novel approach for the representation of the PDL mechanical behaviour.

In order to model long term OTM and not only initial tooth mobility, a remodelling

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algorithm has to be integrated into the mechanical behaviour of the bone, using here a biomechanical approach, focusing on the mechanics and considering remodelling on its phenomenological aspects. For the past decade, there has been a profusion of models, either constitutive laws or remodelling algorithms, for trabecular bone at a macroscopic level. On the one hand, the constitutive laws for trabecular bone account for morphological effects, including anisotropy through the use of fabric tensors, as well as non-linear materials effects, such as (anisotropic) plasticity or damage. However, these complex models usually assume a constant morphology, such as in [3]. On the other hand, remodelling algorithms usually assume small strain linear elasticity for the bone tissue, such as in [4]. We propose here to reconcile both these approaches with a constitutive model for trabecular bone at macroscopic level, built on morphological parameters such as the fabric tensor, accounting for effects such as plasticity of the trabeculae, that can be used in finite strains, and for which the continuum parameters such as stiffness can evolve with morphology as remodelling occurs in the tissue.

3. MODEL

3.1 Geometry and periodontal ligament representation.

A geometry was obtained from the INRIA/GAMMA repository[4], consisting of a surface reconstruction of the mandible (no differentiation between cortical and trabecular bone available) and its teeth (crown and root). The 2D outline in the mesiodistal plane of the left central incisor was extracted and meshed (so that it is composed of a finite number of nodes). However, the PDL is not present in the geometry. Indeed, it cannot be extracted from CT data. We therefore need first to find a way to represent this periodontal ligament. For this, three FE models were created: one for an actual PDL creation (referred to as the reference model), one for a spring representation of the PDL, and one for a contact representation. We then first tested the initial mechanical response of these three models to show we can use contact conditions for the periodontal ligament. For this, material behaviours are first assumed to be linear isotropic (see Table 1). For the spring model, the spring stiffness takes into account the PDL Young's modulus and the distance between consecutive springs. Some non linearity is accounted for as the springs initial length is zero while the PDL thickness is 0.2 mm. For the contact model, bilateral sticking contact conditions allows for an inter-penetration of half the thickness of the PDL and a penalty factor accounting for the PDL Young's modulus is used. We also compare nonlinear material models between the spring representation and the contact representation of the PDL, accounting for its stiffness dependence on the strain as in [5]. A pressure representative of a 1N force was applied on the labial side of the tooth crown and the basal line of the bone was considered fixed.

Table 1: Material parameters for the comparative study.

	Young's modulus [MPa]	**Poisson's ratio [-]**
Bone	1770.	0.3
Tooth	20230.	0.3
PDL	0.6	0.45

An FE analyses was then performed on each model using Metafor[5]. No remodelling is

[4] http://www-roc.inria.fr/gamma/gamma.php
[5] http://metafor.ltas.ulg.ac.be

present at this stage as we only want to compare the PDL mechanical representation.

3.2 Alveolar bone material model, including remodelling

As stated in the introduction, the alveolar bone remodelling is the key factor of OTM. Therefore, its modelling is also of major importance. Alveolar bone is composed of an anisotropic arrangement of trabeculae, here considered as an elastoplastic material (von-Mises plasticity). The remodelling phenomenon mainly leads to a change both in the density of the trabecular network and its orientation. A remodelling model integrated in the constitutive law is here presented. We use as constitutive law a validated model at the tissue level representing the anisotropic arrangement of the trabeculae through the use of a fabric tensor [6]. The fabric tensor is translated into an anisotropic damage-like tensor; the damage is therefore representative of the morphology, not the actual damage of the trabeculae. The remodelling is considered as having an effect on the fabric tensor, therefore modelled with an evolution law. Following other models [7,8], the fabric tensor variation is assumed to be proportional to the difference between a measure of the strain energy density at trabecular level (referred to as the remodelling stimulus) and a homeostatic value of the same measure. In OTM, it is assumed that the pressure level in the PDL (through the activity of fibroblasts) is the key stimulus to differentiate between formation and resorption [9]. As we represent the PDL as non linear contact conditions at the bone/tooth interface, we do not have access to the pressure level in the PDL. However, we here translate the hydrostatic pressure level in the bone into a pressure value in the PDL, using as a scaling factor the bulk modulus ratio of the two tissues. This pressure dependency appears in the remodelling algorithm through a remodelling rate directly function of the hydrostatic pressure (and not only through the strain energy density). We therefore have a remodelling law which can be fully integrated into the constitutive model as it does not depend on any field values of other tissues. In particular, it satisfies the local action principle for a constitutive law in finite strains.

3.3 Two cases of orthodontic tooth movement.

We present here two different cases representative of OTM. The first one is a force driven problem (FDP), similar to the one presented in the comparative study for the PDL model. The tooth is considered as linear elastic; the alveolar bone has an initial apparent density of 50% and is initially considered as isotropic (see Table 2).

Table 2: Material parameters for the remodelling study.

	Young's modulus [MPa]	**Poisson's ratio [-]**
Bone	(Initial value) 1770.	0.3
Tooth	(FDP) 20230	0.3
	(DDP) rigid	-

The second case is a displacement driven problem (DDP): the tooth is rigidly translated (1 mm, i.e. five times the PDL width) in a labio-lingual direction. The displacement prescriptions of the bone are the same as the previous problem.

For both simulations, the maximal force or displacement is applied almost instantaneously (application of the force/displacement in one physical minute) and kept constant over several physical weeks. Remodelling is observed, and leads for the FDP to a modification in the tooth displacement and for the DDP to a modification of the force needed for the applied displacement.

4. RESULTS AND DISCUSSIONS

4.1 Periodontal ligament representation

The results for the tipping movement simulations of the tooth initial mobility produced by the 1N force are presented in Figure 1. Both the linear models with springs or the linear bilateral contact model ensure the transfer of the hydrostatic pressure through the ligament with the same intensity as for the reference model. However, on the lingual side, the shear intensity of the spring model is higher than the reference one and its maximum position is less apical than the reference maximal position.

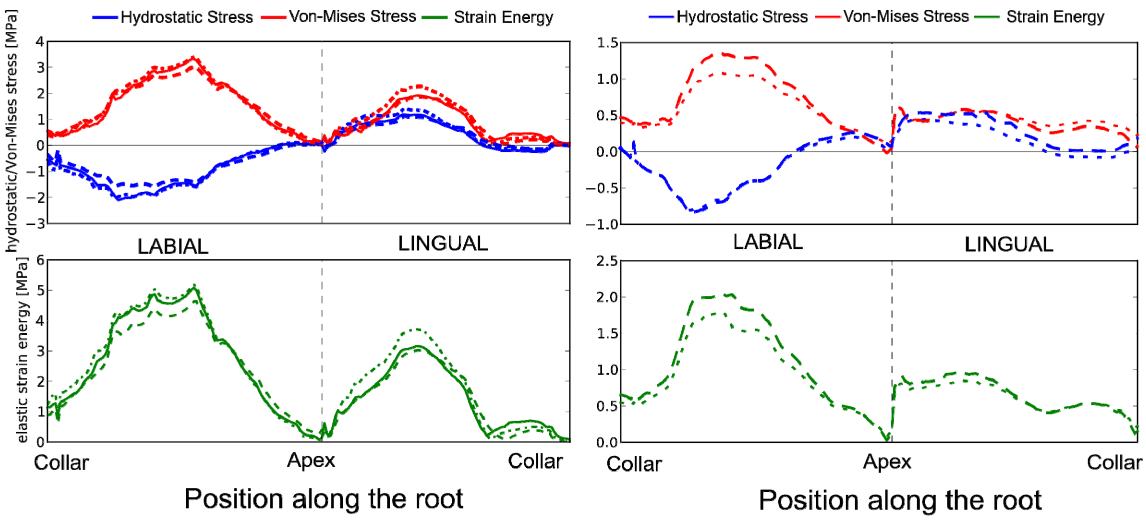


Figure 1: Stresses (top) and strain energy (bottom) in the bone along the root (left: linear model – right: non linear model). Plain lines are obtained for a solid PDL, dotted lines for springs, dashed lines for bilateral contact conditions.

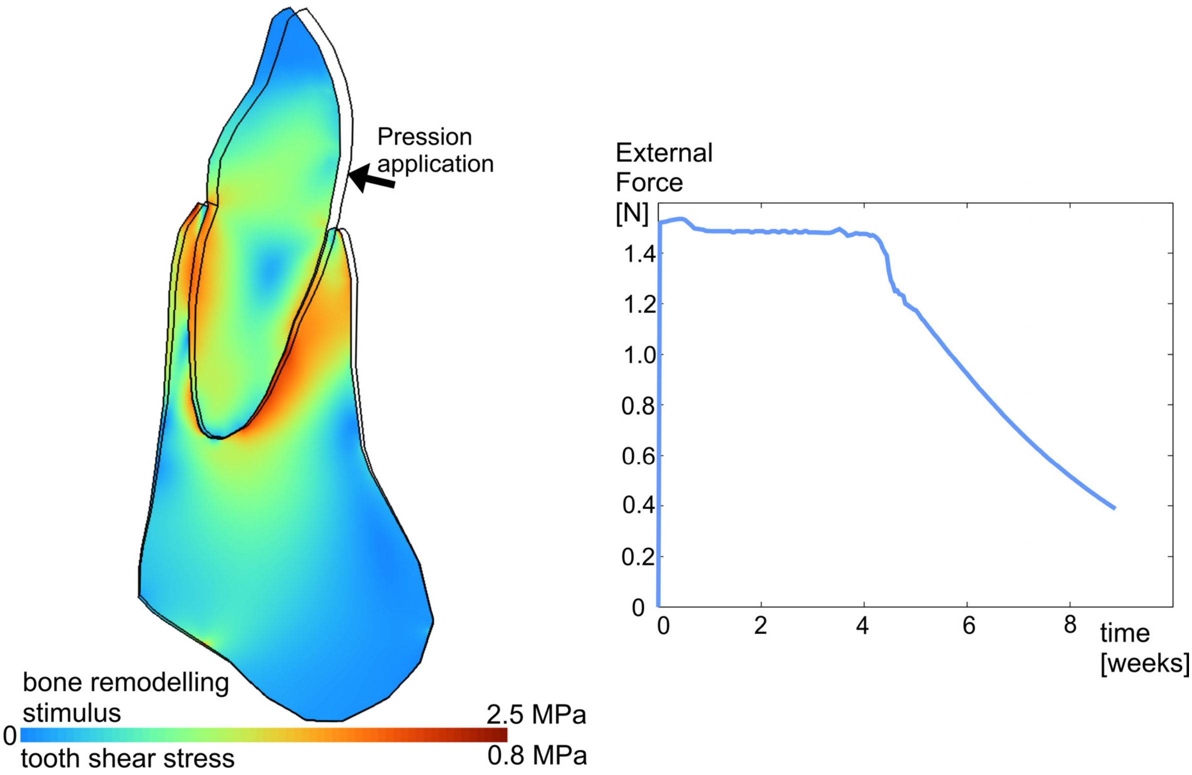
The contact model shows a shear intensity lower than the reference model on the labial side with the same position for the maximal value of shear. For the non linear models, comparison can be made only between spring models and contact models. One can see the shear intensity is higher for the contact model on the labial side while the hydrostatic pressure is higher on the lingual one. The non linear models also show hydrostatic pressure and shear intensities twice as low as the linear models produce. They also lead to a stress intensity distribution which is more homogeneous on the lingual side than the linear models. The non linearity of the material therefore has a large impact on the results. From these observations, we can conclude that using a non linear contact method to represent the periodontal ligament mechanical response gives results on the stress and strain intensity in the bone that are similar to a non linear spring model. Both the spring and the contact models also give similar results in the linear case than the inclusion of a PDL enclosure between the tooth and the bone. Therefore, the contact model will be used from now on as the bone/tooth geometrical interface does not need to be moved to include the PDL and it also does not require conforming meshes at the interface (while the spring model does).

4.2 Force driven problem.

The applied pressure on the crown leads to a rotation movement of the tooth with a centre-of-rotation situated at about one third of the tooth height as can be expected from such a tipping movement. Therefore, the hydrostatic stress distribution is anti-symmetric from this rotation centre. We observe compression at the lingual collar and the labial apex while tension conditions are observed at the labial collar and lingual apex. This leads to two sets of antagonist remodelling sites around the tooth root. Both sites show each on both sides of the tooth a remodelling stimulus higher than the reference stimulus (with a stimulus higher in the compression parts). In classical remodelling theories both sites would therefore undergo bone apposition. In a remodelling model accounting for the fibroblast activity such as proposed, the compression parts of both sites undergo resorption and the traction ones apposition, therefore leading to a tooth rotation as expected. As the applied pressure does not give a pure moment, a slight translation is also observed (see Figure 2(a)).

4.3 Displacement driven problem.

A tooth translation obviously leads to a hydrostatic stress distribution which is more homogeneous than for the previous problem. However, the pressure in the alveolar bone is larger at the apex than at the collar, leading to a slower remodelling as a hyaline zone needs to be resorbed (accounted for in the model when the remodelling stimulus is higher than a maximal physiological level). Once this zone is resorbed, the force needed to keep the fixed displacement decreases slowly to zero after 4 weeks (see Figure 2(b)).



(a) **(b)**

Figure 2: (a) – Remodelling stimulus and von-Mises equivalent stress in the tooth for the FDP after 5 weeks (constant pressure on the crown); **(b)** - Force variation as a function of time for the DDP (fixed translation)

5. CONCLUSIONS

5.1 Geometry and periodontal ligament representation

Clinical CT data precision does not allow for the PDL surface reconstruction. Extensive pre-processing is often used to create a PDL. This study demonstrated the potential of using customized contact conditions on the bone/tooth interface, as both the hydrostatic and shear stress in the bone could be represented while reducing the pre-processing of the model.

5.2 Long term orthodontic tooth movement.

We proposed a remodelling algorithm of the alveolar bone fully integrated to the constitutive law. The pressure dependency of the bone remodelling, mainly due to the fibroblast activation within the PDL, is accounted for considering a remodelling rate directly function of the hydrostatic pressure. This pressure is evaluated in the bone and corrected as an equivalent PDL pressure through a scaling factor representative of the bulk modulus ratio of both materials. Using such a remodelling law allows us to represent long term orthodontic movement, both in force driven and displacement driven problems. In particular, it allows representing the latency period of remodelling, observed between the application of the force and the remodelling phase. Further work should investigate not only other type of boundary conditions due to orthodontic appliances but also 3D models and interactions between the different teeth that are involved in OTM.

6. REFERENCES

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